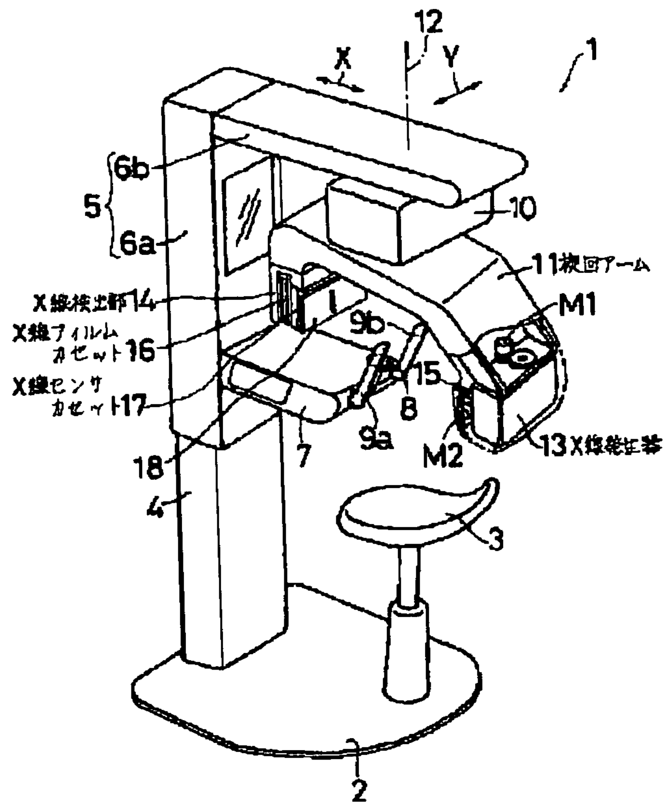
# EUROPEAN PATENT OFFICE

## Patent Abstracts of Japan




PUBLICATION NUMBER : 09135829
PUBLICATION DATE : 27-05-97

APPLICATION DATE : 16-11-95
APPLICATION NUMBER : 07297228

APPLICANT : MORITA MFG CO LTD;

INVENTOR : YASUDA KOJI;

INT.CL. : A61B 6/14 A61B 6/00 A61B 6/00

TITLE : TOMOGRAPH FOR MEDICAL TREATMENT


ABSTRACT : PROBLEM TO BE SOLVED: To shorten a time required for tomography for facilitating and quickening medial examination and treatment by equipping a plurality of different types of X-ray detection means for the selective use thereof.

SOLUTION: A swing arm 11 is mounted on an arm support body 5 fitted to a pillar 4, so as to be capable of swinging around a vertical swing axis 12, and one end of the swing arm 11 is provided with an X-ray generator 13. The other end of the swing arm 11 is provided with an X-ray detection part 14. Also, the X-ray deteaction part 14 has an X-ray film cassette 16 for housing an X-ray film, and an X-ray sensor cassette 17 for housing a CCD sensor or a MOS sensor. In this case, constitution is so made that when one of the cassettes 16 and 17 is selected, the other retreats from a photographing position, thus enabling the selected cassette to be kept at the photographing position.

# A novel semiconductor pixel device and system for X-ray and gamma ray imaging.

David Allison[3], Agamemnon Epenetos[1], Panu Jalas[1,2], Zaheed Karim[5], Melvyn Myers[3], Risto Orava[2], Jouni Pyyhtiä[1,2], Jaakko Salonen[4], Bal Sanghera[3] Miltiadis Sarakinos[2], Tom Schulman[1,2], Konstantinos Spartiotis[1], Ilkka Suni[4], Cao Tieliang[1,2]

[1]Simage Oy, Tekniikantie 12, 02150 Espoo, Finland

[2]SEFT (Research Institute for High Energy Physics in Finland), Siltavuorenpenger 20 C, FIN-00014, Finland

[3]Hammersmith Hospital/Royal Postgraduate Medical School, Department of Diagnostic Radiology, Du Cane Road, London W12 0HS, UK

[4]VTT (Technical Research Center of Finland), Otakaari 7 B, 02150 Espoo, Finland

[5]Hong Kong University of Science and Technology, Department of Electrical and Electronic Engineering, Clear Water Bay, Kowloon, Hong Kong

XP-002207643

P.000-00-1996
P.1248-1250 (3)

*Abstract*

We are presenting clinical images (objects, mammography phantoms, dental phantoms, dead animals) and data from a novel X-ray imaging device and system. The device comprises a pixel semiconductor detector flip-chip joined to an ASIC circuit. CdZnTe and Si pixel detectors with dimensions of the order of 1 $cm^2$ have been implemented with a pixel pitch of 35 $\mu$m. Individual detectors comprise, therefore, tens of thousands of pixels. A novel ASIC accumulates charge created from directly absorbed X-rays impinging on the detector. Each circuit on the ASIC, corresponding to a detector pixel, is capable of accumulating thousands of X-rays in the energy spectrum from a few to hundreds of keV with high efficiency (CdZnTe). Image (X-ray) accumulation times are user controlled and range from just a few to hundreds of ms. Image frame updates are also user controlled and can be provided as fast as every 20 ms, thus offering the possibility of real time imaging. The total thickness of an individual imaging tile including the mounting support does not exceed 4 mm.

Individual imaging tiles are combined in a mosaic providing an imaging system with any desired shape and useful active area. The mosaic allows for cost effective replacement of individual tiles. A scanning system, allows for elimination, in the final image, of any inactive space between the imaging tiles without use of software interpolation techniques.

The Si version of our system has an MTF of 20% at 14 lp/mm and the CdZnTe version an MTF of 15% at 10 lp/mm.

Our digital imaging devices and systems are intended for use in X-ray and gamma-ray imaging for medical diagnosis in a variety of applications ranging from conventional projection X-ray imaging and mammography to fluoroscopy and CT scanning. Similarly, the technology is intended for use in non destructive testing, product quality control and real time on-line monitoring. The advantages over existing X-ray digital imaging modalities (such as digital imaging plates, scintillating screens coupled to CCDs etc.) include compactness, direct X-ray conversion to an immediate real time digital display, exquisite image resolution, dose reduction and large continuous imaging areas.

Our measurements and images confirm that this new digital imaging system compares favourably to photoluminescence plates and photographic film.

## I. Introduction

The need to replace film radiography by filmless approaches is widely acknowledged in the medical community. Replacing an all purpose medium used for image acquisition, storage and display, with a technology optimising each task will result in productivity improvements of radiology departments: real time imaging, elimination of consumables (film, chemicals) and tasks (film handling), facilitation of image display, archiving and transfer are some of the advantages [2].

By reducing the number of faulty exposures and consequent repetitions, a digital technology will reduce average patient received dose [1]. The film's limited dynamic range (2 orders of magnitude) should be replaced by a device offering a linear response over several orders of magnitude. A medium with higher radiation detective efficiency will result in better Signal-to-Noise ratios (S/N) or, where S/N is noncritical, to dose reductions. At the same time, the good spatial resolution of the film should not be compromised.

The desirability of digital radiography has led to the development of various modalities. Current state of the art involves the digitisation of film images, the Digital Imaging Plate (DIP) using the phenomenon of photostimulated luminescence, or real time imaging modalities involving Selenium based detectors [2], amorphous silicon [3] [4] [5] [6] and Charged Coupled Devices coupled to scintillating screens [7]. Gaseous wire chamber designs have also been proposed [8] [9] [10].

We have developed a semiconductor pixel device for high intensity X-ray imaging. Our design is based on the direct conversion of X-rays and can be implemented with a number of semiconductor devices of which we are implementing and testing Si and CdZnTe. Instead of opting for large monolithic imaging area, a highly impractical approach due to high cost and low yield, we have developed a tiling method joining monolithic detectors (tiles) into large area mosaics with no inactive regions, a convenient and cost-effective approach. Convenience and cost effectiveness are also provided by the minimisation of digital electronics.

We provide the first preclinical and clinical data using this novel device and compare its performance with other digital methods and standard film.

## II. DESCRIPTION

Figure 1 depicts our device comprising a pixel semiconductor substrate.

A detailed description of our system can be found in [11].

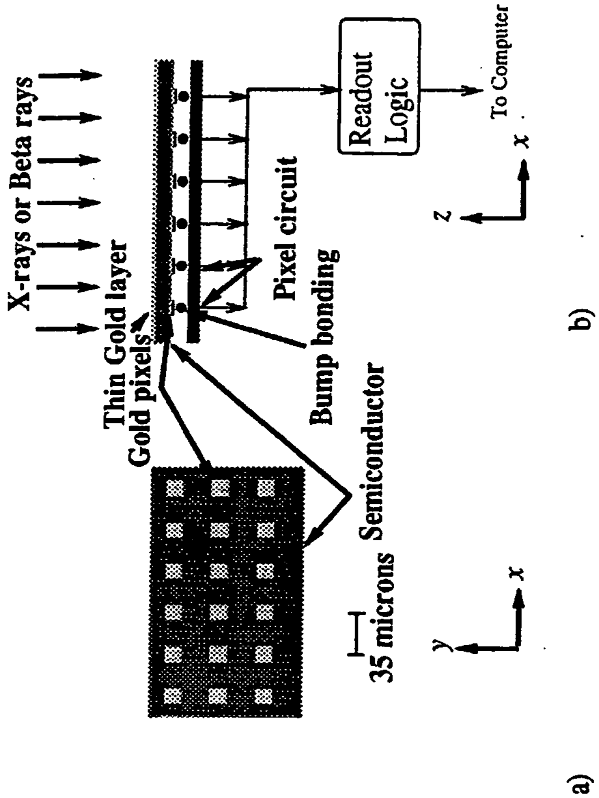



Fig. 1 Our basic setup is shown in (b). Simage design includes the readout logic. Beyond the readout logic the information goes to a computer for image reconstruction, display and storage. (a) shows only a magnified view of the semiconductor surface. We illustrate the gold pixels which can be seen in an attached photo. This is the smallest pixel pitch ever achieved.

Pixels are patterned with photolithography [12] and the pixel pitch is 35 $\mu$m. The CdZnTe surface is passivated in order to ensure stability and high surface resistivity. Interpixel resistivities of 200 - 1000 G$\Omega$ have been measured. Each detector pixel is indium bump-bonded to an Application Specific Integrated Circuit (ASIC). ASICs are read out sequentially by the readout logic and in cycles.

The imaging area of a monolithic detector is $12.2 \times 4.2$ mm$^2$ containing 41,000 pixels. Detector thicknesses may vary from 0.3 mm to 3 mm. For our tests we have chosen 0.3 mm thickness for Si and 1.5 mm for CdZnTe. For CdZnTe this thickness translates to an absorption efficiency of practically 100% for photons up to 60 keV. Worth noting is that photon absorption in CdZnTe is predominantly photoelectric; hence, the ionisation from each incident photon is confined within a region small compared to the pixel size.

Incident radiation generates charge which is accumulated at the pixel ASIC. The integrated front end amplifiers operate in the charge integration mode with a charge storing capacity of more than 16 million electrons. The integrated charge is converted to a current signal which is multiplexed at 2 - 5 MHz to a 12 bit ADC. The pixels remain active during multiplexing enabling multiple image frame acquisition and real time imaging.

Each pixel circuit is readout sequentially by the readout logic. While the pixel circuits of a row are readout, the circuits of the previous row are cleared. For our tested monolithic detector, this amounts to an inefficiency due to inactive time of 0.4%.

We operate Si and CdZnTe detectors at bias voltages of 60 V and 350 V respectively.

We choose exposure times (single exposure) in the range of 10 - 100 ms and operate with an X-ray tube (Siemens Polymobil III) capable of being operated at voltages from 40 kV to 100 kV and currents up to 2 mA.

Our device offers high dynamic range: greater than 3 orders of magnitude for a single exposure, unlimited for real time motion imaging (fluoroscopy when device used with video processing). Exposure times are user selected and may vary from 10 ms to several seconds depending on the application.

Individual detectors are tiled together on a master plane to form a larger imaging area [13] The tiles are mounted non-destructively on the master plane and can easily be removed and replaced. Thus the same master plane can be used for both Si and CdZnTe detectors. The inactive space between the tiles is completely eliminated by moving the master plane. The imaging tiles are arranged in shifted columns enabling the whole image area to be covered with image snap shots at three different positions of the master plane. With an appropriate step motor driven translation stage the total imaging time is less than 800 ms.

## III. RESULTS

The modulation transfer function (MTF) has been measured for the Si detectors to be 20% at the Nyquist frequency of 14 lp/mm. An MTF of 15% at 10 lp/mm was measured with a CdZnTe detector (see Figure 2).

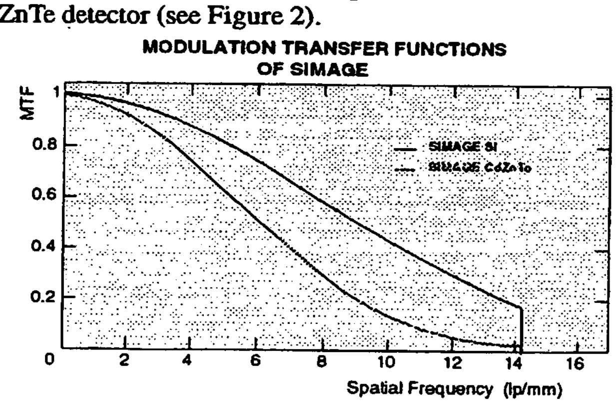



Fig. 2 The measured Modulation Transfer Functions of our system. The curves for both Silicon and CdZnTe are shown.

The MTF was measured by obtaining the detector response

to a thin slit (10 $\mu$m) aligned with a pixel row.

Comparative results from imaging a Leeds test phantom (TORMAS with 36mm scatter material) demonstrate a comparative advantage of our device with respect to other methods at lower doses. Both phosphor and film images suffer a loss of position resolution as the mAs is decreased but the CdZnTe image still results in 7.1 lp/mm.

A mouse paw was imaged using the same imaging modalities and the CdZnTe image was seen to show improved contrast.

We obtained images of various test objects: a diode, a potentiometer an integrated chip and a human finger.

We have tested a multitile scanning system by imaging a complete mouse. The functionality of the method has been demonstrated.

## IV. CONCLUSION

The functionality of a digital X-ray imaging system based on flip chip joined CdZnTe and Si pixel detectors has been demonstrated. Acquired X-ray images show superior quality over existing digital and film imaging systems. Future development of the system will focus on large area imaging in medical applications utilising the demonstrated tiling method.

## V. REFERENCES


[1] W.Krampla et.al. "Clinical experience in a digital radiology department," *Österreichische Krankenhaus-Zeitung 34(1993) Special Issue on Radiology.*

[2] Denny L. Lee et.al. "A new digital detector for projection radiography," *SPIE Medical Imaging Conference, Feb. 1995*

[3] Gene P. Weckler "Large Area Imager Development," *9th International Workshop on Room Temperature Semiconductor X-ray and G-ray, Associated Electronics and Applications, Sept. 18-22, 1995, Grenoble, France.*

[4] J.Chabbal et.al. "Amorphous Silicon X-ray Image Sensor," *SPIE Symposium, Medical Imaging 1996, 10-15 Feb, 1996, Newport Beach, USA*

[5] EG&G Reticon "MX0256 Amorphous Silicon Imaging Array," Advance Information

[6] Thorsten Graeve et.al. "Amorphous silicon image sensor for X-ray applications," *SPIE Proceedings on Charge-Coupled Devices and Solid State Optical Sensors V, Vol 2415, 1995*

[7] H.Liu et.al., "Lesion Detectability Considerations for an Optically-coupled CCD X-ray Imaging System," *IEEE Trans. in Nucl. Sci. Vol. 41 No.4. August 1994*

[8] S.E.Baru et.al., "Digital X-ray Imaging Installation for Medical Diagnostics," *Nuclear Instruments and Methods in Physics Research A238 (1985)* pp. 165-169.

[9] G. Charpak " Will gaseous detectors become routine tools in biology and medicine?," *Nuclear Instruments and Methods in Physics Research A283 (1989)* pp. 371-374

[10] F. Sauli "Applications of gaseous detectors in astrophysics, medicine and biology," *Nuclear Instruments and Methods in Physics Research A323(1992)* pp. 1-11

[11] GB patent 9508294.7 granted, PCT/EP95/02056

[12] GB patent 9524387.7 pending

[13] GB patent 9605978.7 pending, PCT EP96/03559